

Finite Element Analysis of Fixed-bearing Unicompartamental Knee Arthroplasty: Coronal Alignment of the Tibial Component under Physiological Alignment

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3 **Alignment of the Tibial Component under**
4 **Physiological Alignment**

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Abstract

Background: To investigate the influence of tibial component varus and valgus angles on knee joint biomechanics in fixed-bearing unicompartmental knee arthroplasty (FB-UKA) with physiological mechanical alignment of the lower limb.

Methods: A normal knee joint model was established and validated. Nine simulation conditions were created based on the physiological mechanical alignment of the lower limb, with the tibial component positioned at neutral (0°), and varus/valgus angles of 3° , 6° , 9° , and 12° in the coronal plane. The peak von Mises stress and maximum principal strain were analyzed across different regions, including the superior surface of the polyethylene (PE) bearing, the cartilage surface of the lateral compartment, and the cancellous and cortical bone surfaces beneath the tibial component.

Result: In the FB-UKA model, a tibial component valgus angle exceeding 3° significantly increases the peak von Mises stress and maximum principal strain on the anteromedial proximal cortex. During tibial component varus alignment, changes in the peak von Mises stress and maximum principal strain within the cancellous bone are initially minimal; however, as the varus angle continues to increase, stress levels approach the yield threshold, which may lead to bone resorption and prosthetic loosening. Furthermore, a further increase in the valgus angle exacerbates the load on the contralateral compartment, potentially accelerating the progression of contralateral osteoarthritis.

Conclusion: This study demonstrates that in FB-UKA models with physiological limb mechanical alignment, static analysis results are significantly influenced by the position of the tibial component. We recommend that, as long as the global hip-knee-ankle (HKA) is not

malaligned, the coronal alignment of the tibial component can be safely targeted between neutral and mild varus. This strategy may help reduce the risk of complications and improve implant survivorship.

Keywords

Unicompartmental Knee Arthroplasty; Finite Element Analysis; Biomechanics ; Knee Osteoarthritis

1. Introduction

The incidence of Knee Osteoarthritis (KOA) is very high among people over 50 years old (1). Approximately one-third of patients present with isolated medial compartment involvement in the early stages (2). Unicompartmental knee arthroplasty (UKA) has been in clinical use since the 1970s. Since its introduction, its application for unicompartmental KOA has been a subject of intense debate. The studies by Newman et al. and Sershon et al. found no significant differences between UKA and Total Knee Arthroplasty (TKA) in early opioid use, return-to-work time, long-term functional outcomes, or revision rates. However, UKA offers superior early postoperative knee range of motion, less surgical trauma, and a shorter hospital stay (3, 4). Therefore, UKA is undoubtedly a more suitable option for patients with unicompartmental KOA.

Some studies have indicated that the reasons for revision after UKA may be closely associated with various postoperative complications, such as anterior knee pain, aseptic loosening of the prosthesis, polyethylene bearing wear, and progression of lateral compartment osteoarthritis (5). Compared with TKA, UKA differs significantly in its surgical techniques and underlying principles. During UKA, the integrity of the cruciate ligaments is preserved. Furthermore, rather than aiming for a standardized mechanical alignment of the lower limb, UKA seeks to restore the patient's unique native alignment while correcting varus or valgus deformities. Kennedy et al. conducted a retrospective study on the clinical outcomes and postoperative alignment of 100 cases

of medial UKA. The results indicated that the clinical effect was more ideal when the mechanical axis was located in the central area of the knee joint or slightly medial(6). In a clinical study reviewed by Slaven et al. involving 3,351 cases, patients with well-functioning UKAs at 10-year follow-up averaged mild varus mechanical alignment of approximately 4° (7). A study by Gill et al. demonstrated that, in FB-UKA, knees with mild varus tibiofemoral alignment achieved better clinical scores than those with valgus alignment(8). While the overall lower limb alignment is crucial, the coronal alignment of the components plays an equally critical role. Regarding the coronal alignment of the femoral component. Finite element analysis (FEA) studies by Ma et al. and Wang et al. suggested that the femoral prosthesis should be aligned with the center of the distal femoral condyles to reduce medial and lateral translation(9, 10). However, recommendations for the coronal alignment of the tibial component remain controversial. While Nie et al. and Koh et al. advocate placing the tibial component in a neutral to 3° varus position, and recommend avoiding valgus placement (11-14). Conversely, Innocenti et al. and Kang et al. proposed neutral alignment or less than 3° of valgus (15, 16). However, consensus has not been reached. The research by Iesaka and Sawatari et al (17, 18), contrarily suggests that the tibial prosthesis should be 2° valgus to 4° valgus. Previous studies have largely assumed a neutral mechanical axis, leading to contradictory conclusions and highlighting the necessity of conducting research under physiological alignment. Therefore, based on the studies by Kennedy and Slave et al. FEA was employed to construct a 4° varus FB-UKA model ($HKA = 4^{\circ}$, Figure 1), simulating a common physiological variation. Within this controlled environment, the femoral component was maintained in a neutral position, while the coronal angle of the tibial component was systematically adjusted as the independent variable. This study aims to evaluate the effects of varying tibial component angles on the peak von Mises stress and maximum principal strain in the tibial cortical bone, cancellous bone, polyethylene (PE) bearing, and lateral compartment cartilage, thereby clarifying the biomechanical significance of tibial component positioning under a 4° physiological varus alignment.

Fig. 1 Hip-knee-ankle angle (HKA)、 joint line obliquity (JLO)、 lateral distal femoral angle (LDFA)、 medial proximal tibial angle (MPTA).

2.Materials and methods

2.1 Native knee model

A healthy male volunteer (approved by the ethics committee and having provided informed consent) with no history of knee disease or trauma was recruited. The volunteer's knee joint was scanned in the supine position using a 128-slice CT scanner (Siemens Medical, Erlangen, Germany; slice thickness: 0.6 mm) and a 3.0 T MRI system (GE Signa HDxT, USA; slice thickness:1 mm). The scanning range covered 15 cm of both the distal femur and the proximal tibia, and the acquired data were saved in DICOM format. The DICOM data were imported into MIMICS 21.0 (Materialise NV, Leuven, Belgium) to reconstruct the cortical bone and cancellous bone based on specific CT thresholds. Concurrently, the menisci, cartilage, and ligaments were reconstructed using the MRI data. The masks for each structure were refined using the Draw and Erase tools within the Edit Masks module of Mimics to adjust contours and fill internal regions. These reconstructed geometries were then exported in STL (Stereolithography) format. Subsequently, the 3D models were corrected in Geomagic 2018 (Geomagic, North Carolina, USA) using the Smoothing, Triangle Reduction, and Remesh modules to generate smooth bone structures, and the files were saved as IGES format. Finally, the assembly and solid model construction were completed using SolidWorks 2020 (Dassault Systèmes S.A., USA).

2.2 UKA knee model

Based on the obtained knee model, a size 5 fixed-bearing unicompartmental knee arthroplasty (FB-UKA) prosthesis (JOURNEY Unicompartmental Knee System, Smith & Nephew, USA; provided by the manufacturer) was selected. The femoral component had an anteroposterior (AP) diameter of 52 mm and a mediolateral (ML) diameter of 23 mm, yielding a ratio of approximately 2.26. The tibial component had an AP diameter of 52.3 mm and a ML diameter of 30.4 mm, with a ratio of approximately 1.72. The thickness of the polyethylene bearing is 6mm. These components were then imported into SolidWorks 2020. According to the surgical standards, Boolean operations were used to simulate the UKA surgery on the medial knee joint. Coronal and sagittal sectioning ensured that the femoral component was positioned at the center of the medial femoral

condyle, covering the central and distal posterior aspects of the medial femoral condyle as comprehensively as possible (19), while avoiding varus/valgus angulation, medial/lateral translation, and axial rotation during the simulated UKA. According to previous studies, the tibial posterior slope was set to 7° (20, 21) to align with the patient's native tibial anatomy. In the coronal plane, the tibial component was positioned perpendicular to the femoral component with no axial rotation. The tibial component was placed with maximal coverage, and its edges were fully supported by the surrounding cortical bone rim. Houskamp et al. demonstrated that tibial bone stock should be preserved as much as possible. They found that a resection depth exceeding 5.82 mm leads to a significant decrease in the mean failure load of the tibia, thereby increasing the risk of periprosthetic fractures. Furthermore, excessive resection often necessitates the use of thicker polyethylene inserts, which may elevate the long-term risk of aseptic loosening. Based on these considerations, we set the tibial plateau resection thickness to 6 mm in our model (22-24). A standard FB-UKA model was established by appropriately adjusting the joint line height and setting the HKA angle to 4° of varus, while avoiding excessive stretching of the medial collateral ligaments (MCL) and lateral collateral ligaments (LCL). We established a right-handed Cartesian coordinate system was established, centered at the geometric center of the tibial plateau. The axes were oriented as follows: the Y-axis aligned with the tibial mechanical axis (proximal direction), the X-axis pointed medially, and the Z-axis pointed anteriorly. To simulate coronal plane alignment, varus and valgus rotations were applied as rotational moments around the Z-axis (anteroposterior axis), perpendicular to the mechanical axis. The tibial component was set with a 4° varus inclination relative to the lower limb mechanical axis. The tibial component alignment was referenced to the lower limb mechanical axis with a 4° varus orientation, while maintaining the joint line height consistent with the standard anatomical position. A series of three-dimensional (3D) FEA models were established, including 12° , 9° , 6° , and 3° varus, as well as 3° , 6° , 9° , and 12° valgus (Figure 2). The models were saved as STP files.

Fig. 2 FB-UKA models with different varus and valgus angles: (A-D) = varus 3° – 12° , (E-H) = valgus 3° – 12°

The models were then imported into ANSYS 21.0 (ANSYS, Inc., USA) for meshing, material

property assignment, and boundary condition setting. All structures were meshed using 10-node tetrahedral elements. Convergence was subsequently verified, defined as a change of greater than 5% between adjacent meshes. The average element size for the simulated cartilage and menisci was 0.8 mm (25). The experimental process is shown in Figure 3.

Fig. 3 Flow chart of finite element analysis.

According to previous studies, the ligaments were defined as hyperelastic isotropic materials and represented by the neo-Hookean model. The constitutive equation is defined as $\Psi = C_1(I_1 - 3)$ where C_1 represents the initial shear modulus and I_1 denotes the first invariant of the right Cauchy-Green deformation tensor. The C_1 values for the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), MCL, LCL were set to 5.83 MPa, 6.06 MPa, 6.43 MPa, and 6.06 MPa, respectively (26). The remaining structures were defined as linear elastic isotropic materials (16, 26, 27). The specific material parameters for each structure are detailed in Table 1 (28-32). Six contact pairs of femoral and tibial cartilage, tibial cartilage and meniscus, and femoral cartilage and meniscus were established. Using the penalty function algorithm (33). Tie constraints were applied between the ligaments, cartilage, and bone, as well as between the tibial cartilage and meniscus. The interactions between the femoral cartilage and meniscus, and between the femoral cartilage and tibial cartilage, were defined as frictionless finite sliding surface-to-surface contacts. This contact formulation ensures that the stability of the knee joint is maintained by the structural equilibrium state rather than by friction (33). The boundary conditions were set at the relative extension position of the femur and tibia. Specifically, only the flexion-extension motion of the femur was constrained, while motions in other directions were left free. The distal ends of the tibia and fibula were fully constraint (34).

Tab. 1 Properties of various structural materials

Component name	Modulus of elasticity (Mpa)	Poisson's ratio
Cortical bone	17000	0.3
Cancellous bone	350	0.25
Meniscus	27.5	0.33
Cartilage	15	0.46
Cobalt-chromium-molybdenum alloy	210000	0.3
Polyethylene bearing	850	0.4

2.3 Stress loading and model validation

All UKA models employed the same stress loading protocol. A 1000 N load was applied vertically downward along the femoral mechanical axis. The boundary conditions remained unchanged and were held constant during the loading process. Convergence was verified after loading (27).

The normal knee model was validated using two methods (32, 35, 36), Method 1: A 1000 N load was applied vertically downward along the femoral mechanical axis. The distribution, peak values, and load ratio of the equivalent stress on the cartilage of the medial and lateral compartments were observed. Method 2: An anterior drawer test was simulated. The boundary conditions were modified to restrain only the flexion-extension motion of the tibiofibular axis, while the motion of the femur was fully restrained. A 134 N anterior load was applied perpendicular to the coronal plane at the attachment site of the anterior cruciate ligament (ACL). The anterior tibial displacement was measured. The results of the above experiments were compared with those of previous studies. If the differences were minimal, the model was deemed valid.

3.Result

3.1. Verification results of model validity

Validation of the normal knee finite element model: The normal knee joint model can converge after the load is applied. The verification results of Method 1 show that the peak von Mises stress of the medial meniscus and cartilage of the tibia is 7.66 Mpa, and that of the lateral meniscus and cartilage is 5.71MPa. The load ratios of the inner and outer chambers were 55.3% and 44.7% respectively (Figure 4) . These studies were similar to previous ones (32, 37, 38). The verification results of Method 2 show that the forward displacement distance of the anterior side of the tibial loading point is 4.79mm, which is similar with the standard displacement of approximately 5 mm in the anterior drawer experiment (32, 39).

Finite element analysis of UKA and normal knee models under identical boundary conditions: When the tibial component was positioned in the standard alignment, the load distribution between the medial and lateral compartments of the FB-UKA model was evaluated. The medial compartment bore 58.9% of the load, while the lateral compartment bore 41.1%, yielding a load ratio of approximately 3:2. This finding is consistent with the study by Innocenti et al (15). Furthermore, a distinct stress concentration was observed in the anteromedial tibial cortex, indicating an increase in stress in this region postoperatively. Conversely, due to stress shielding, the Von Mises stress in the anteromedial tibial cancellous bone decreased, these results are consistent with the findings of Sawatari et al (40-42). The areas of stress concentration were primarily located at the junction of the horizontal and vertical tibial resection planes and around the tibial tray keel. Additionally, no significant changes in the cancellous bone stress of the lateral plateau were observed when compared to the normal knee model.

The 3D finite element models of the normal knee and FB-UKA developed in this study showed good agreement with previous validation results, indicating high reliability. Therefore, they are suitable for further biomechanical analysis. The numbers of elements and nodes in the UKA model are summarized in Table 2.

Tab.2 The number of units and nodes of models with different varus and valgus angles

Category	Number of units	Number of nodes
valgus12°	353075	623971
valgus9°	344802	609562
valgus6°	344636	608298
valgus3°	322622	543721
0°	324231	546215
varus3°	345265	589634
varus6°	321598	583652
varus9°	351249	619784
varus12°	409787	722357

Fig. 4 Peak von Mises stress and its distribution on the tibial cartilage and meniscus surfaces. (A) Medial tibial cartilage and medial meniscus; (B) Lateral tibial cartilage and lateral meniscus.

3.2 Polyethylene bearing

When the tibial component was in the standard position (0°), the peak von Mises stress on the PE bearing was 21.53 MPa. At varus angles of 3°, 6°, 9°, and 12°, the peak von Mises stress were 29.38, 35.20, 46.43, and 72.51 MPa, respectively; at valgus angles of 3°, 6°, 9°, and 12°, the peak von Mises stress were 33.59, 50.24, 74.41, and 89.23 MPa, respectively. Similarly, in the standard position, the maximum principal strain (compressive strain) of the PE bearing was 37,032 $\mu\epsilon$. At 3° of varus and valgus, the maximum principal strain increased by 18.75% (43,977 $\mu\epsilon$) and 62.64% (60,230 $\mu\epsilon$), respectively. As the angle increased, the maximum principal strain increased by 157.41% (95,323 $\mu\epsilon$) at 12° of varus and by 242.87% (126,970 $\mu\epsilon$) at 12° of valgus. The peak von Mises stress and maximum principal strain in the medial compartment of the PE insert increased significantly with increasing varus and valgus angles, with a greater magnitude of change observed under valgus conditions. Under varus conditions, the peak von Mises stress and maximum principal strain were concentrated in the central annular area of the insert, whereas under valgus conditions, these regions shifted transversely (Figures 5 and 6).

Fig. 5 Peak von Mises stress distribution on the polyethylene bearing: (A-D) = valgus 12°–3°,

(E-H) = varus 3°–12°

Fig. 6 Peak von Mises stress and maximum principal strain of the polyethylene bearing in the FB-UKA model under different varus and valgus alignments

3.3 Tibial cortical bone

In the standard position, the peak von Mises stress in the tibial cortical bone was 27.86 MPa. At varus angles of 3°, 6°, 9°, and 12°, the peak von Mises stress were 29.34, 32.20, 51.84, and 55.90 MPa, respectively; at valgus angles of 3°, 6°, 9°, and 12°, the peak von Mises stress were 37.60, 48.52, 56.10, and 59.55 MPa, respectively. In the standard position, the maximum principal strain in the medial tibial cortical bone was 1679 $\mu\epsilon$. This increased to 3733 $\mu\epsilon$ (a 122.33% increase) at 12° of varus, and to 3540 $\mu\epsilon$ (a 110.84% increase) at 12° of valgus. Similar to the PE insert, the peak von Mises stress and maximum principal strain in the medial tibial cortical bone increased significantly with increasing varus and valgus angles. Notably, at 6° of varus, the maximum principal strain increased at a faster rate compared to 6° of valgus (Figure 7).

Fig. 7 Peak von Mises stress and maximum principal strain of the tibial cortical bone of FB-UKA models under various varus and valgus alignments

3.4 Tibial cancellous bone

Under standard neutral alignment, the peak von Mises stress in the medial tibial cancellous bone was 1.77 MPa. At varus angles of 3°, 6°, 9°, and 12°, the peak von Mises stress were 2.05, 2.83, 2.98, and 4.52 MPa, respectively; at valgus angles of 3°, 6°, 9°, and 12°, they were 2.50, 2.89, 3.90, and 5.66 MPa, respectively. Similarly, the maximum principal strain in the medial tibial cancellous bone at 0° was 5069 $\mu\epsilon$. This increased to 14859 $\mu\epsilon$ (a 193.23% increase) at 12° of varus, and to 16996 $\mu\epsilon$ (a 235.29% increase) at 12° of valgus. Due to the dual-keel design of the tibial component, supplemented by a medial keel ridge structure, the peak von Mises stress and maximum principal strain were primarily concentrated between the medial and lateral keels of the tibial component. At varus angles less than 9°, the maximum principal strain increased at a slower rate compared to valgus angles (Figure 8).

Fig. 8 Peak von Mises stress and maximum principal strain of the tibial cancellous bone of FB-UKA models under various varus and valgus alignments

3.5 Lateral compartment cartilage

At the neutral position, the peak von Mises stress on the cartilage surface of the lateral compartment was 2.11 MPa. At varus angles of 3°, 6°, 9°, and 12°, the stresses were 1.59, 1.55, 1.50, and 1.17 MPa, respectively; at valgus angles of 3°, 6°, 9°, and 12°, they were 2.33, 2.47, 2.80, and 3.69 MPa, respectively. The maximum principal strain on the cartilage surface of the lateral compartment at the neutral position was 171,060 $\mu\epsilon$. This decreased to 82,514 $\mu\epsilon$ (a 51.76% reduction) at 12° of varus, and increased to 319,380 $\mu\epsilon$ (an 86.71% increase) at 12° of valgus. Unlike other observation targets, the peak von Mises stress and maximum principal strain in the lateral compartment cartilage sequentially decreased from 12° of valgus to 12° of varus (Figure 9).

Fig. 9 Peak von Mises stress and maximum principal strain of the lateral compartment in FB-UKA models under various varus and valgus alignments

4. Discussion

In the treatment of medial unicompartmental KOA, UKA offers significant advantages over TKA by maximizing the preservation of bone and cartilage tissue, and providing superior postoperative knee function (43). However, the limited surgical exposure in UKA significantly increases the risk of intraoperative component malalignment, particularly in obese patients. A study by Van der et al. (44) indicated that the primary causes of revision in fixed-bearing UKA (FB-UKA) are: progression of osteoarthritis in the contralateral compartment (36%), aseptic component loosening (28%), and polyethylene liner wear and joint instability (12% each).

Specifically, the progression of disease in the contralateral compartment is primarily

292 attributed to overcorrection of the mechanical axis during surgery, or to an overreliance on stress
293 radiographs without comprehensive CT or MRI evaluations, which leads to the failure to detect
294 occult preoperative bone and cartilage damage. Conversely, component loosening and wear are
295 mainly caused by abnormal stress distribution at the bone-implant interface, which accelerates
296 implant failure and reduces long-term survivorship (45-47). The alignment of the knee prosthesis
297 is a key factor influencing its biomechanical performance and longevity. Based on the studies by
298 Kennedy and Slave et al. a finite element model of UKA with varying coronal alignments of the
299 tibial component was established using a 4° varus HKA angle as the baseline. The aim was to
300 determine the optimal range for the coronal alignment of the tibial component through a
301 comprehensive analysis of peak von Mises stress and maximum principal strain. In recent years,
302 the Constitutional Proximal Articular geometry of the Knee (CPAK) classification has gained
303 widespread attention for its utility in evaluating coronal knee morphology and revealing individual
304 alignment variability. Given that the most common CPAK types I and II in Asian populations
305 predominantly exhibit a constitutional varus phenotype (48, 49), selecting a 4° varus baseline in
306 this study is well-supported by population-specific anatomical evidence. Although this model
307 primarily focuses on assessing the biomechanical effects of component alignment relative to this
308 physiological baseline, surgeons should still consider patient-specific CPAK phenotypes in
309 clinical practice to achieve optimal joint line restoration.

310 This study found that the stress on the polyethylene bearing was significantly higher than that
311 on the articular cartilage of the lateral compartment, and the load proportion in the medial
312 compartment was also greater. This stems from the difference in material properties. The
313 polyethylene liner is made of high-molecular-weight polyethylene, whereas the meniscus and

cartilage are composed of chondrocytes, collagen fibers, glycoproteins, and water(50),The elastic modulus of cartilage is much lower than that of the polyethylene insert. Under the same pressure, cartilage can deform to convert part of the stress into potential energy, thereby reducing local stress. The geometry and conformity of the PE bearing are critical determinants of contact stress and implant longevity. FEA by Koh et al. (51)demonstrated that conformity significantly influences stress distribution. For an anatomically designed UKA, the tibial insert and the lateral meniscus exhibited the lowest contact stress during the stance phase of the gait cycle. However, during the swing phase, this anatomical UKA design resulted in the highest contact stress on both the tibial insert and the lateral meniscus. In contrast, for flat-type UKA designs, the articular cartilage exhibited the lowest contact stress under both gait and squat loading conditions. Studies by Kwon et al. (52)have shown that mobile-bearing designs generally exhibit lower articular contact pressure than fixed-bearing inserts, as the mobile platform allows for self-alignment, thereby maximizing the contact area. However, mobile bearings may also present with pronounced backside stress at the inferior surface, potentially increasing the risk of wear. In contrast, fixed-bearing designs can sometimes transfer higher stresses to the contralateral compartment. A follow-up study by Lim et al. (53), with a mean follow-up of 10.6 years, found no statistically significant differences in WOMAC scores, range of motion, or radiographic outcomes between the fixed-bearing and mobile-bearing groups. Nevertheless, the 10-year survivorship free from revision was superior for fixed-bearing UKA (97.8%) compared to mobile-bearing UKA (88.9%). In our current model, we selected a flat-type fixed-bearing PE insert to achieve an optimal balance among these biomechanical factors. When the polyethylene bearing is in the standard position, the peak von Mises stress of the insert is minimized. As the varus or valgus

angle increases, the peak von Mises stress also increases and exhibits a symmetrical distribution. Under valgus alignment, the peak von Mises stress in the insert exhibited substantial fluctuations, with an increase of more than 50% across all models. In contrast, under varus alignment, the changes in peak von Mises stress were relatively gradual, accompanied by a more uniform stress distribution and an increase of approximately 20%. This discrepancy may be attributed to the fixation mechanism beneath the tibial component. The medial keel increases the contact area between the implant and the cancellous bone. Given the lower elastic modulus of cancellous bone, it can undergo deformation to accommodate changes in varus angles, thereby buffering the variation in peak von Mises stress.

The study results indicate that the peak von Mises stress and maximum principal strain in the tibial cortical bone are primarily distributed across the anteromedial aspect of the tibial plateau. In previous studies, Iesaka and Sawatari et al. (17, 18) observed significant changes in stress on the anteromedial surface of the tibia as the varus angle increased. They recommended that the tibial component be positioned with a valgus angle between 2° and 4°. Similarly, Simpson et al. (41) demonstrated in their study on Oxford UKA that cortical bone stress increases sharply when the varus angle exceeds 8°. Our results are consistent with these previous studies under various conditions. Therefore, for the specific prosthetic design in this study, an increase in the valgus angle induces a redistribution of stress. Compared to cancellous bone, the elastic modulus of cortical bone is hundreds of times greater; consequently, the maximum principal strain in the cortical bone remains relatively low, with the majority of stress in the medial compartment being borne by the cortical bone. Poor matching of the UKA tibial component can lead to various complications: component overhang may cause soft-tissue impingement and postoperative pain, whereas insufficient coverage of the resected bone surface increases the risk of component loosening and subsidence(54). Because the bone support area of a UKA tibial component is less than half of that in TKA,

precise sizing is critical. Studies by Carpenter and Fitzpatrick et al.(55, 56) have shown that commercially designed implants fail to achieve satisfactory cortical bone coverage; even with specifically designed prostheses (with coverage rates of 77% on the medial side and 60% on the lateral side), complete coverage cannot be attained. Furthermore, Lee et al.(57) found that when the tibial component cannot fully match the tibial plateau, the uncovered areas of the central medial and posterior lateral cortices are relatively safe. However, to effectively prevent component subsidence, it is imperative to ensure adequate coverage of the posterior medial and anterior lateral aspects of the tibial plateau. As the valgus angle increases, the load-bearing center of the tibial component shifts laterally, subjecting the anteromedial tibial cortex to high stress and strain. This exacerbates microfractures and adaptive bone remodeling, leading to the formation of sclerotic bone, which represents one of the causes of anteromedial knee pain.

In revision UKA surgery, aseptic loosening of the tibial component accounts for approximately one-third of cases (44). From a clinical perspective, several factors can increase the risk of tibial component loosening. First, an overly aggressive initial tibial resection may compromise the structural integrity of the proximal tibia. Furthermore, patient-specific factors play a critical role; for instance, postmenopausal women typically exhibit poorer cortical bone quality or osteopenia, which reduces the structural support for the implant. Additionally, obesity imposes a higher mechanical load on UKA, which can further exacerbate the risk of implant subsidence when cortical coverage is inadequate(23, 57). Michel et al. (58) reported that the yield stress of tibial cancellous bone is approximately 3.57 MPa. When stress levels approach this threshold, bone damage is further exacerbated, potentially leading to prosthetic loosening. Brumby et al. (59) reported that in UKA, the multiple drill holes created for guide pins and the keel of the tibial component compromise the compressive strength of the proximal tibia. This predisposes the bone to fractures originating from these holes, typically occurring between 3 and 18 weeks postoperatively (median, 8 weeks). Furthermore, the use of peripheral pegs that violate the medial tibial cortex should be strictly avoided, as this can even precipitate tibial plateau fractures. Our

results indicate that the peak von Mises stress exceeded the yield stress of cancellous bone only in the 9° valgus and 12° valgus models. However, even at 6° of varus/valgus, the peak von Mises stress reached 80% of the yield limit, indicating a high-risk critical state. Pegg and Chang (42, 60) found that the peak von Mises stress occur at the intersection of the tibial horizontal and vertical bone resection planes. In contrast, in our study, the peak von Mises stress and maximum principal strain was primarily concentrated around the two porous regions. This discrepancy may be due to the single-keel dual-column structure of the tibial prosthesis used in this study. The maximum principal strain (compressive strain) around the keel exceeding 4000 $\mu\epsilon$ can lead to microfractures and fatigue in the adjacent cancellous bone, thereby increasing the risk of loosening. This finding is consistent with the study by Taylor (31). To avoid excessive tibial resection and the associated complications, Chang et al(60). proposed a novel osteotomy technique that adjusts the orthogonal geometry to a radial configuration. This approach effectively reduces local stress and prevents degenerative remodeling, providing crucial guidance for clinical surgical practice.

On the cartilage surface of the lateral compartment, the peak von Mises stress and maximum principal strain gradually decreased from 12° of valgus to 12° of varus, which is consistent with the findings of Zhu et al. (61). Throughout the knee joint, the structures of the medial compartment can be regarded as an integrated unit. As the valgus angle decreases or the varus angle increases, the load gradually shifts toward the medial compartment, resulting in a corresponding reduction in the load proportion on the lateral side. Excessive valgus alignment can lead to a significant increase in the peak von Mises stress and maximum principal strain in the lateral compartment, thereby accelerating the progression of lateral compartment osteoarthritis.

This study established a structurally intact knee joint finite element model based on CT and MRI images. Finite element analysis has been widely applied and developed in orthopedic biomechanical research due to its advantages of high precision, low cost, and good repeatability. However, this study has several limitations: First, although performing the sensitivity analysis using a single volunteer eliminates inter-subject variability, it also limits the representativeness of the results. Patients with KOA exhibit altered joint mechanics and various anatomical adaptations; therefore, finite element models reconstructed from healthy volunteers cannot fully capture the

anatomical alterations present in KOA patients. Future studies should incorporate patient-specific KOA models to validate these findings and explore models representing diverse anatomical structures and various prostheses. Second, our analysis was limited to the static stance condition. However, human knee motion is inherently a dynamic process, and analyzing models under different gait conditions might yield different conclusions. Future research should incorporate musculoskeletal modeling to validate these findings under dynamic conditions. This will be addressed in our subsequent studies. Third, we restricted rotational activity of the knee joint when applying loads to simplify the stress distribution test. This aspect will be further evaluated through future in vitro mechanical testing. There are always differences between models and clinical cases; model constraints may also include factors such as age, gender, weight, and race. Future research can establish boundary conditions and loading scenarios closer to real-world conditions to further advance this investigation.

5. Conclusion

Our findings indicate that in FB-UKA it is biomechanically safe to position the tibial component in mild varus, provided that the overall physiological limb alignment is maintained. Within this range, the peak von Mises stress and maximum principal strain across knee components remained within physiological limits, showing no significant increase. This suggests that moderate varus alignment of the tibial component does not compromise implant survivorship or increase the risk of complications. Therefore, we recommend that, as long as the global HKA is not malaligned, the coronal alignment of the tibial component can be safely targeted between neutral and mild varus.

Data availability statement

The original contributions presented in the study are included in the article material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by Ethics Review Committee for Scientific Research Projects of Zhongshan Hospital Affiliated to Dalian University. Ethics Number: KY2023-019-1; In accordance with the principles of the Helsinki Declaration of 1964 (revised in 2013) all collected clinical and imaging data are known to patients, The participants provided their written informed consent to participate in this study.

Consent for publication

Not Applicable.

Conflict of interest

The authors declare that they have no competing interests.

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Author contributions

BW: software, visualization, formal analysis, methodology, data curation, writing—First author.
YN: formal analysis, methodology, data curation. CY: formal analysis, methodology, data curation. JF: formal analysis, methodology, data curation. JW、LG and LX: methodology, data curation, writing — Review and editing. SW: methodology, resources, data curation —

Corresponding author.

All authors contributed to the article and approved the submitted version.

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This study was led by BW, with YN, CY, and JF assisting in data analysis and methodology development. JW, LG, and LX were responsible for reviewing and editing the manuscript, while SW, as the corresponding author, provided resources and guidance. We appreciate the professional contributions of all authors to this research. This research was supported by the 2025 Henan Province Medical Science and Technology Key Project (LHGJ20250505).

Clinical trial number

Not applicable

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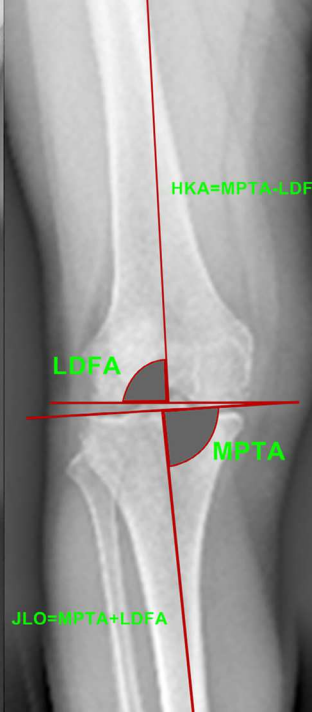
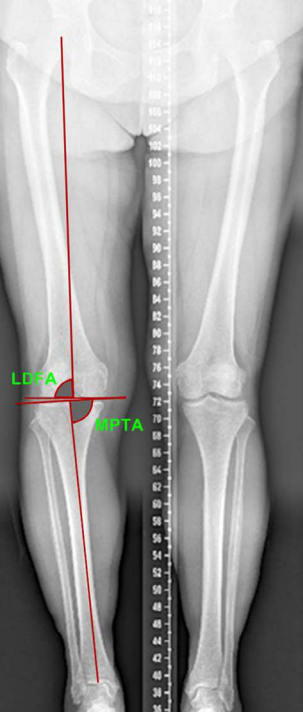
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List of abbreviations

FB-UKA	Fixed-bearing unicompartmental knee arthroplasty
KOA	Knee Osteoarthritis
TKA	Total Knee Arthroplasty
UKA	Unicompartmental Knee Arthroplasty
FEA	Finite Element Analysis





A



B



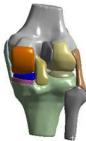
C



D



E



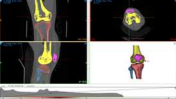
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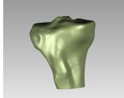
G



H



Extract a rough model in
Mimics



Smooth the model in
Geomagic



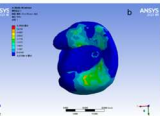
Assemble in
SolidWorks



Perform surface fitting
and solid modeling for
the UKA implant

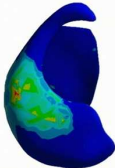


Import into Ansys for
mechanical analysis



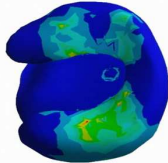
Analysis of Data Results

A: Stress and Strain
Stress-effect
Model: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



A

A: Stress and Strain
Stress 2
Effective (von-Mises) Stress
Unit: MPa
Time: 1



B

A: -7
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



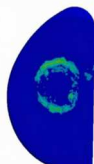
A

A: -8
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



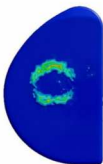
B

A: -6
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



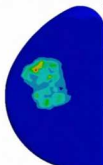
C

A: -5
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



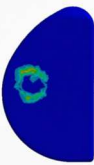
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A: -9
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



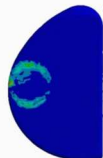
E

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Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



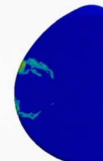
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A: 6
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



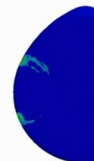
G

A: 9
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



H

A: 12
Equivalent Stress 4
Type: Equivalent (von-Mises) Stress
Unit: MPa
Time: 1



I

